

Detecting of the manifestation of myocardium anisotropy by the magnetocardiographic method

V. Sosnytsky¹, M. Budnyk¹, I. Chaikovsky², Yu. Minov¹, P. Sutkovyy¹, B. Hailer³, and F. Steinberg²

¹*SQUID AG*; ²*Institute for BioSciences, University Witten/Herdecke*;

³*Philippusstift, Kath. Krankenhaus, Essen, Germany*

1 Introduction

Over the last years, highly sensitive magnetic field sensors, SQUIDS, have been employed to measure the magnetocardiogram (MCG), which are generated by ionic currents flowing in myocardium when it is active.

The MCG method provides additional information in comparison to surface electric potential distribution. The obtained information is complementary in nature because the magnetic field is related to the curl of the impressed current sources into the human heart while the electric field is related to its divergence [1]. It is known that the myocardium has many kinds of anisotropy. One of them is determined by an anisotropy of conductivity (uniform anisotropy), which plays a key role in realistic propagation of the electrical excitation in the myocardium. This was indicated by computer simulation of normal MCG based on a propagation model of an excitation wave front in three-dimensional arrays of cardiac cells within the cellular automata theory [2]. Animal studies and theoretical calculations suggest that magnetic recordings can provide new information about the electrical anisotropy of cardiac tissue [3,4,5] and the problem only exists to reveal and use in clinical practice.

The uniform anisotropy is due to two properties, which are microscopic in nature; the first being the difference in electrical conductivity in the axial and transverse direction of the fibers. Such anisotropy is described by the bidomain theory and characterized by the common fiber orientation defined by global angles in standard spherical co-ordinates [6]. This theory explains experimental potential distributions measured from isolated canine hearts [7] and it was found that the conduction velocity parallel to the fiber axis was greater (3:1) than that of the transverse. Secondly, uniform anisotropy is caused by directional differences in cell-to-cell coupling on a macroscopic level. This anisotropy is described within the framework of the models of discontinuous conduction [8].

Today ECG plays an important role in clinical diagnostics and cardiac research, and an equivalent electric dipole is used as the model of the cardiac electric source. The said dipole, in fact, is the vector of the electro-motive force (EMF), i.e. electric potential, which is also called the electric heart vector (EHV).

The most used dipole within the MCG-method model is an equivalent magnetic dipole which is also called as magnetic heart vector (MHV) and equivalent current dipole (ECD) which is the vector of the impressed current. It was shown that angle between EMF and MHV at the QRS peak is very close to 90° [9]. It should be mentioned that if this angle is exactly equal to 90°, there would be no new information in the MCG compared to ECG. However, it has been found that this angle varies considerably during QRS complex, both from patient to patient [10] and within various cardiac disorders [11].

The above-mentioned results show that the angle between MHV and EHV should contain new information about the myocardium conductivity.

However, in our opinion MHV does not reflect completely electrophysiological phenomena because the physiological sense only has the multipole expansion of the cardiac generator [12]. It is sum of the equivalent current generators-multipoles where ECD is the equivalent current generator of the lowest order.

In the overall anisotropic case a myocardium conductivity is a tensor [13]. However, at the axial symmetry with respect to fiber direction one can consider only two components - transversal ($S_{\perp}$) and axial (S_0) conductivity. The present study is based on the statement that the degree of anisotropic conductivity manifestation is characterized by an angle γ between a direction (α) of a EMF vector and direction (β) of maximal current density (MCD) vector in the human heart at the same time

$$\gamma = \alpha - \beta. \quad (1)$$

Then, isotropic $\mathbf{I}_0$ and anisotropic $\mathbf{I}_{\perp}$ components of ECD vector $\mathbf{J}$ take the form

$$I_0 = J \cos \gamma, \quad I_{\perp} = J \sin \gamma, \\ \tan \gamma = S_{\perp} / S_0. \quad (2)$$

Within the bidomain theory [3] one can identify the I_0 as isotropic current dipole and $I_{\perp}$ as that of anisotropy.

The aim of this study was to investigate the connection between the degree of anisotropic conductivity manifestation (ACM) and myocardial ischaemia in patients with coronary artery disease at normal resting ECG.

2 Methods and Results

We investigated 27 patients, with CAD (stenosis >75% in ≥ 1 vessel), stable angina, without myocardial infarction and with normal resting ECG or unspecific changes of ventricular repolarisation. The control group consisted of 18 healthy volunteers.

The direction α of a vector EMF was calculated at the R- and T-peaks by standard ECG data.

The MCG was registered at a rectangle grid of 6×6 with 4 cm pitch by means of the single-channel magnetocardiographic system (MCG1, SQUID AG, Germany), installed in an unshielded location. The above device provides the registration of the vertical magnetic field gradient component d^2B/dz^2 and reference ECG. The said signals continuously sampled under the discretization frequency of 1000 Hz over a period of 30 s at each point of a grid.

The software provided sampling, filtration, artifact removing, averaging and magnetic field distribution maps calculations.

The program CURLINES realizes the method of magnetic moments (MMM[14]. According to this method a layer of magnetic dipoles distributed on the investigated surface is considered as heart biomagnetic field sources. In this case this surface is a plane located inside a thorax. Using the MMM-method, the correspondence between the unknown distribution of layer density in the sources plane and the distribution of a magnetic field induction of heart received in the measurement plane, is described by appropriate integral equations. The solution of these equations results in a distribution of a magnetic charges double layer density (or, in other words, function of a current), the level lines of which present lines of a current.

The reconstruction of these currents is displayed in the forms of current density maps and current line maps. Current lines are curves along that the current flows, i.e. at any point the elementary current directs tangentially above line. The number (spatial density)

of current lines is proportional to current density at that point. The spatial point of the maximum current density and direction of vector, which is tangential to current line at that point, are determined.

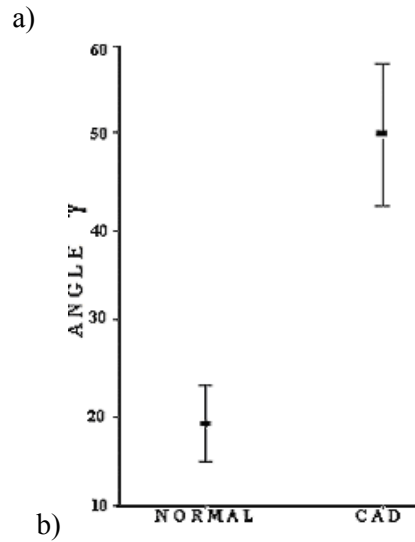
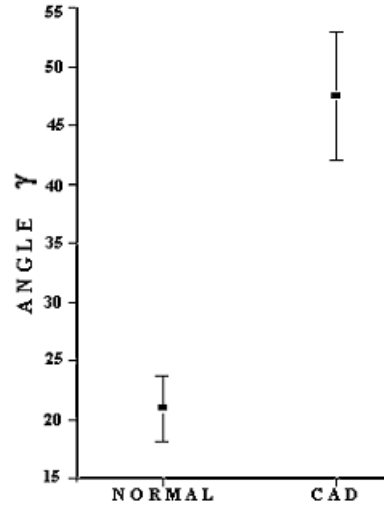


Figure 1: Mean $\pm$ SE values of angles γ : a) R peak, b) T peak.

The vector reflects the direction of the maximum current density (MCD) used for calculation of the β -angle at the R- and T-peaks. After calculation of the β - and α -angles, the value of ACM was obtained by expression (1). From Fig.1 one can see that at the R peak the angle γ lies within the range 20.9 ± 2.82 (mean healthy volunteers, and within the range 47.52 ± 5.45 for CAD patients. Within the T-wave, the angle lies within ranges 19.25 ± 3.84 and 49.8 ± 7.2 for the healthy group and CAD pts, respectively. Thus, the above results suggest that angle of ACM increases in CAD patients in comparison to healthy volunteers. The above

increase is slightly more for T peaks (by almost 2.38 times) compared to the R peak (almost 2.27 times). From the Figure it is clearly seen that the ACM angles for both groups are significantly different. T-test analyses show both groups are significantly different; $p < 0.025$ for the R peak and $p < 0.01$ for the T-wave. Sensitivity of this index is 93%, specificity 85%.

3 Discussion

The effects in the presence of both spiral and uniform anisotropies will be summarized. In the simplest case it leads to simple summation of respective partial angles of ACM (1). The healthy person is characterized by the summary angle (between the EMF and MCD) which are determined by both the spiral structure of the heart and the conductivity anisotropy. Spiral anisotropy changes from one human to another and depends only on anatomical peculiarities. The conductivity changes in a healthy person but it also changes due to heart disorders. The ECG method is insensitive to spiral anisotropy and weakly sensitive to uniform anisotropy, but the MCG method is sensitive to both kinds of myocardium anisotropy. Results [13, 15, 16] suggest that the myocardium anisotropy cannot only be recognized by ECG.

The presence of pathological variations in part from all numbers of myocytes, leads to a change in the average longitudinal component S_0 of the myocardium conductivity. Therefore, disturbances of the excitation wave propagation will be characterized by changing the ACM angle. Thus, increasing the ACM angle (Fig.1) is explained by decreasing of S_0 due to existence of lower conductivity zones under CAD.

As propagation of the above wavefront has two qualitatively different phases, de- and repolarization, the comparison between normal and pts groups at R- and T-peaks was logically performed. However, the use of R and T-peaks are not restricted in the present approach. The calculation of the ACM angle can be performed for any cardiocycle timepoint. The difference between the ACM angle at R- and T-peaks could be explained based on the difference of the excitation front propagation in the course of depolarization (mainly radial) and repolarisation (mainly tangential). The second reason may be that the above phases are characterized by different types of ions mainly carrying the transmembrane currents (i.e. Ca ions for de- and K ones for repolarization). In this case, ACM may be connected with different influences of certain heart disease to conductivity ions of selected species (e.g., due to their mobility or

membrane permeability). Thus, decreasing of the ACM angle at the T-peak in comparison to the R-peak may be explained by increasing the longitudinal conductivity averaged over myocardium involving repolarization at that timepoint. On the other hand, the same effect under CAD is also explained by increasing of the K conductivity or decreasing that of Ca.

4 Conclusion

The proposed approach significantly distinguishes the group of healthy subjects and group of patients with CAD. Thus, analysis based on determination of the ACM using both ECG and MCG data in combination, is potentially useful for identifying CAD with high sensitivity and specificity.

It should be mentioned that the presented approach is oriented towards clinics equipped with very simple ECG and MCG apparatus.

References

1. D. Geselowitz, "The theoretical basis of bioelectromagnetic fields", *Biomedizinische Technik* **42**, 1-4, 1997.
2. B. Horacek, "Digital model for studies in magnetocardiography", *IEEE Trans. Magn.* **Mag-9**, 440-444, 1973.
3. D.J. Staton, R.N. Friedman, and J.P. Wikswo, "High resolution SQUID imaging of octupolar currents in anisotropic cardiac tissue", *IEEE Trans. Appl. Supercond.* **3**, 1934-1936, 1993.
4. B.I. Roth and J.P. Wikswo, "Electrically-silent magnetic fields", *Biophysical J.* **50**, 739-745, 1986.
5. J.P. Wikswo, "Tissue anisotropy the cardiac bi-domain and the virtual cathode effect", in *Cardiac Electrophysiology: From Cell to Bedside*, second Edition, D.P. Zipes and J. Jalife, Eds., Orlando: W.B. Saunders, 1994, pp. 348-361.
6. D. Geselowitz and W. Miller, "A bidomain model for anisotropic cardiac muscle", *Ann. Biomed. Eng.* **11**, 191-206, 1983.
7. P. Colli-Franzone et al, "Potential fields generated by oblique dipole layers – modeling excitation wavefront in the anisotropic myocardium", *Circ. Res.* **51**, 330-346, 1982.
8. W. Ellis et al, "Fractional electrograms from a computer model of heterogeneously uncoupled anisotropic ventricular myocardium", *Circulation* **92**, 1619-1626, 1995.
9. J. Malmivuo and R. Plonsey, "Bioelectromagnetism: Principles and Application of

- Bioelectric and Biomagnetic Fields*", NY-Oxford: Oxford University Press, 1994.
10. J. Nousiainen, J. Lekkala, and J. Malmivuo, "Comparative study of the normal vector MCG and vector ECG", *J. Electrocardiol.* **19**, 275-290, 1986.
 11. J. Nousiainen, "*Behaviour of the vector MCG in normal subjects and in some abnormal cases*", Dr. Tech. Thesis, Tampere Univ. Tech., Tampere, Finland, 1991.
 12. D. Geselowitz, "Multipole representation for an equivalent cardiac generator", *Proc. IRE*, **48**, 75-79, 1960.
 13. S. Broek and M. Peters, "On anisotropic volume conduction", *Biomedizinische Technik* **42**, 21-24, 1997.
 14. S. Romanovych, "Reconstruction of three-components dipoles within layer", *Biomedizinische Technik*, **42**(1), 227-230, 1997.
 15. P. Wach et al, "Cardiac source imaging during de- and repolarization", *Biomedizinische Technik* **42**, 269-272, 1997.
 16. F. Sachse et al, "Comparison of solution to the forward problem in electrophysiology with homogeneous, heterogeneous and anisotropic impedance models", *Biomedizinische Technik* **42**, 277-280. 1997.